(12) UK Patent Application (19) GB (11) 2 184 629 (13) A

(43) Application published 24 Jun 1987

(21) Application No 8629354

(22) Date of filing 9 Dec 1986

(30) Priority data

(31) 8530394 (32) 10 Dec 1985 (33) GB
8600229 7 Jan 1986

(51) INT CL[4]
H04R 25/00

(52) Domestic classification (Edition I)
H4J 30A 30X H

(56) Documents cited

GB	1582821	WO A1	83/02212
WO A1	86/01671	US	4596902
WO A1	85/02085	US	3818149

(58) Field of search
H4J
Selected US specifications from IPC sub-class H04R

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(54) Compensation of hearing

(57) A signal processing device for improving a user's hearing ability comprises signal processing means (5) for processing an electrical signal representative of an auditory signal and supplying the processed signal to a transducer (4) to produce an output auditory signal modified to improve a user's hearing ability. The signal processing means includes a filter (6) having a transfer function determined by settable parameters and a programmable memory (7) for storing the filter parameters. The filter parameters are determined by calibration means (2) in dependence on a measured hearing spectrum of a user, the calibration means (2) optimising the assignment of the available filter parameters.

FIG. 1

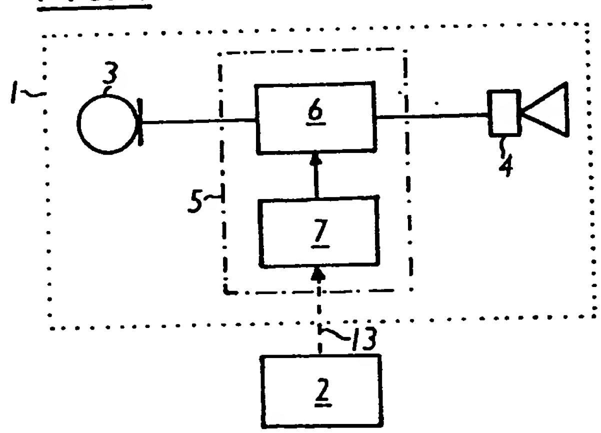

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FIG. 1

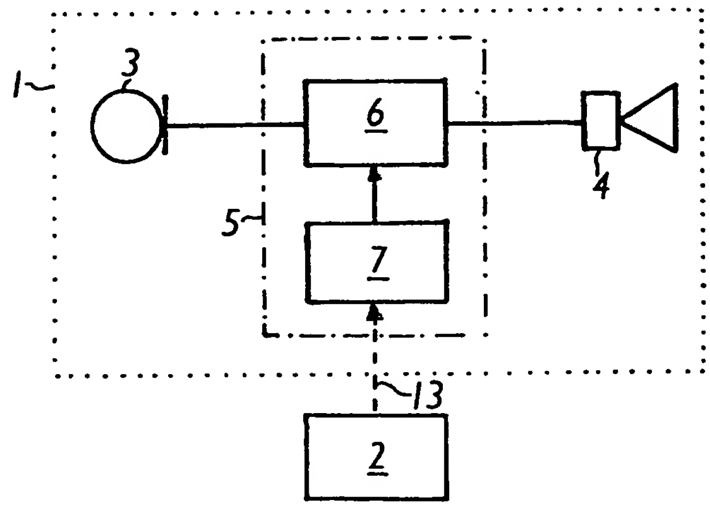

FIG. 2

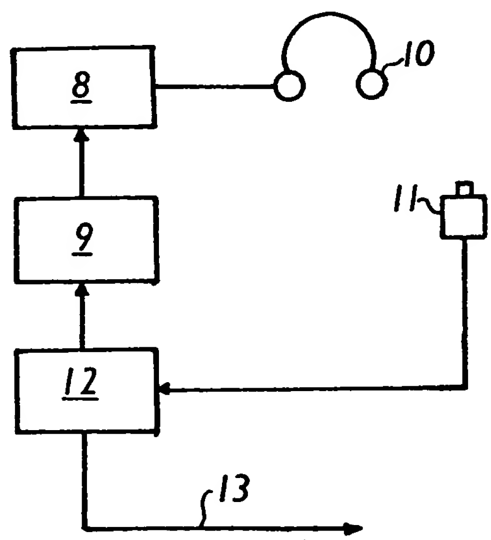

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FIG. 3

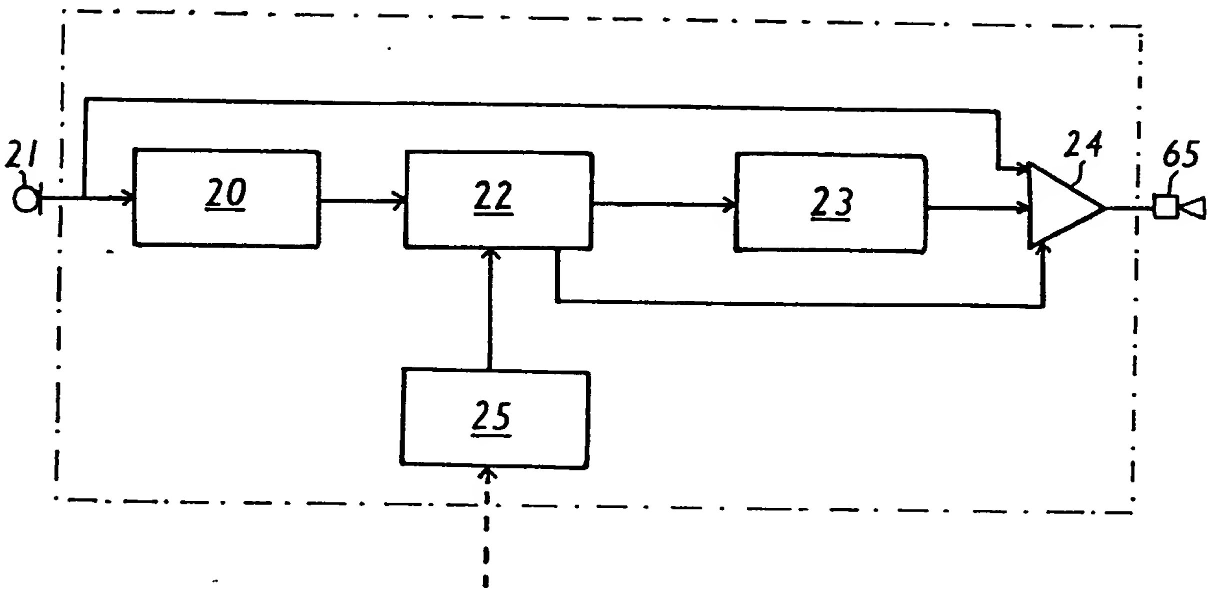
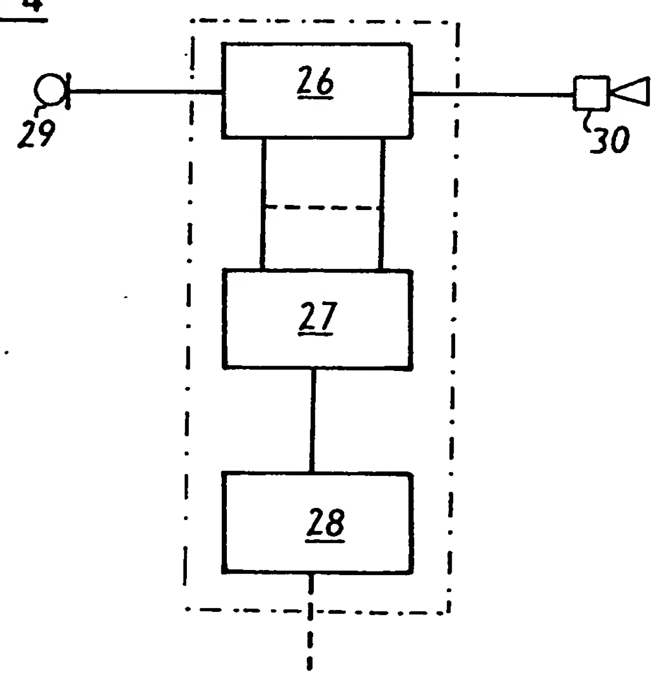


FIG. 4



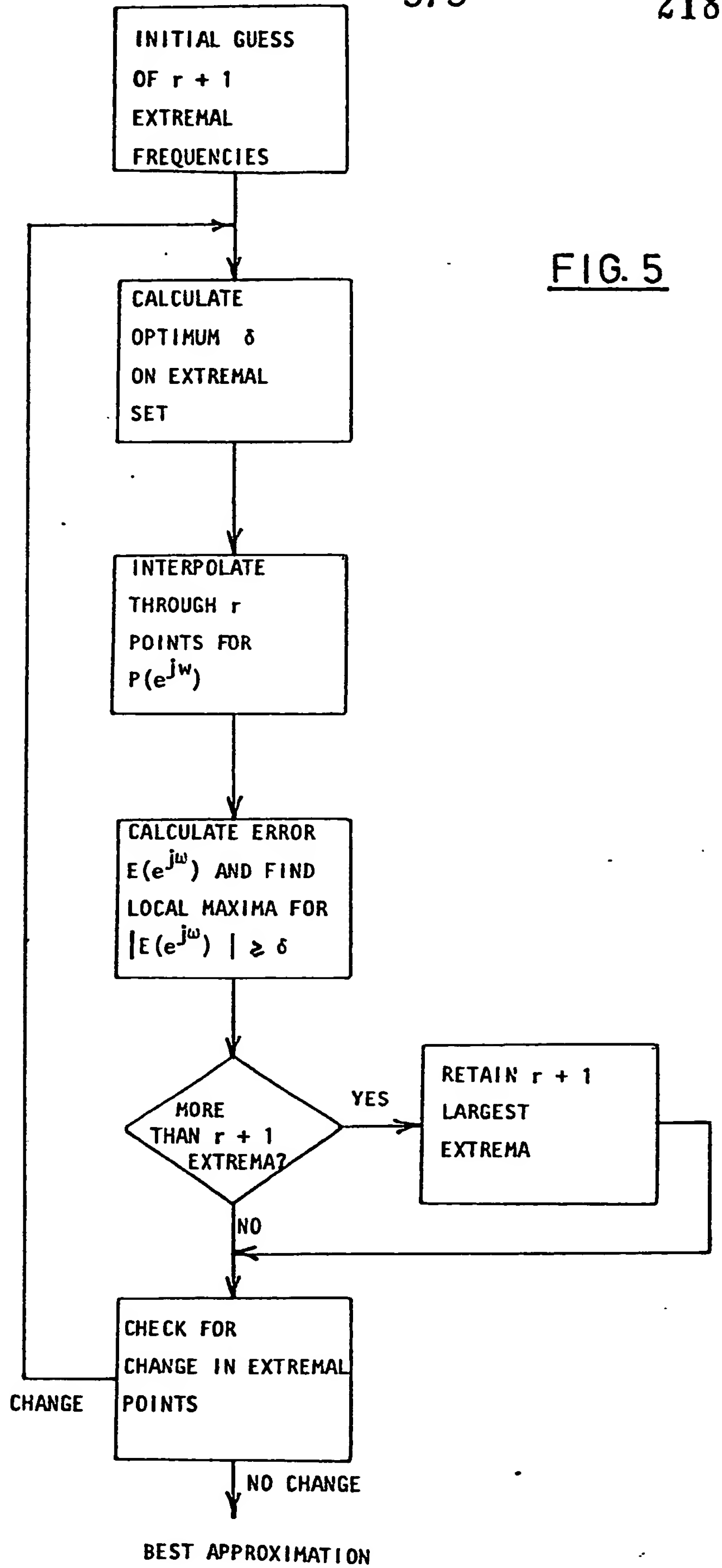

INITIAL GUESS OF r + 1 EXTREMAL FREQUENCIES
CALCULATE OPTIMUM δ ON EXTREMAL SET
INTERPOLATE THROUGH r POINTS FOR P(e^jw)
CALCULATE ERROR E(e^jω) AND FIND LOCAL MAXIMA FOR |E(e^jω)| ≥ δ
MORE THAN r + 1 EXTREMA?
YES
RETAIN r + 1 LARGEST EXTREMA
NO
CHECK FOR CHANGE IN EXTREMAL POINTS
CHANGE
NO CHANGE
BEST APPROXIMATION

FIG. 5

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FIG. 6

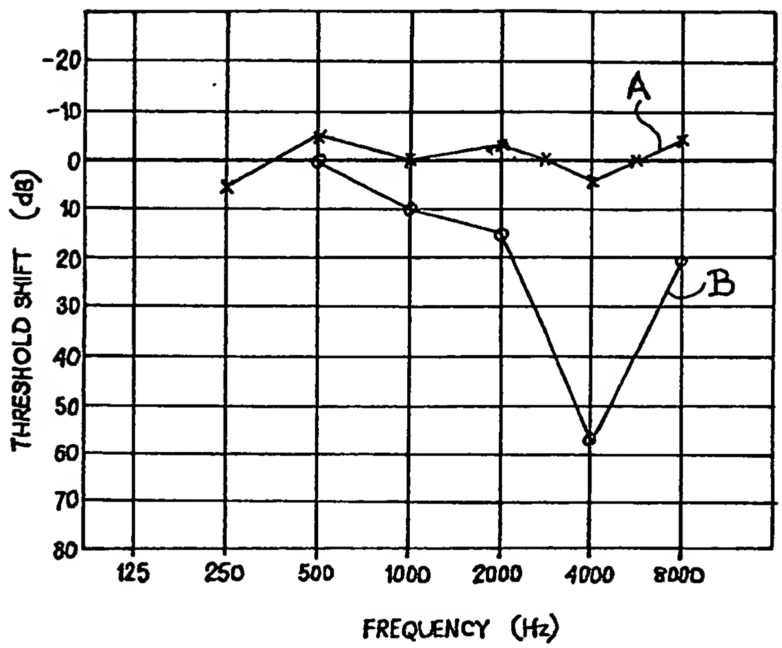

FIG. 7

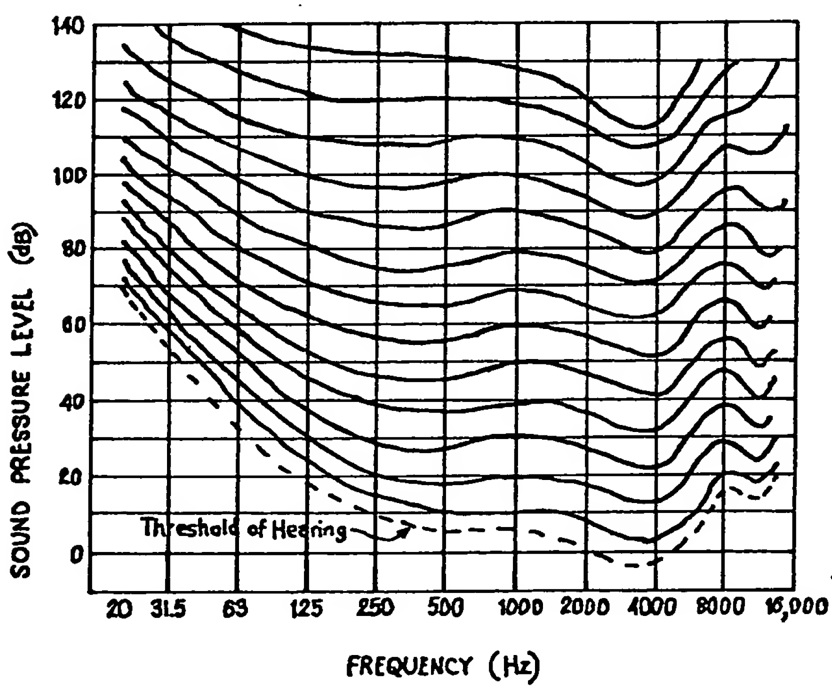

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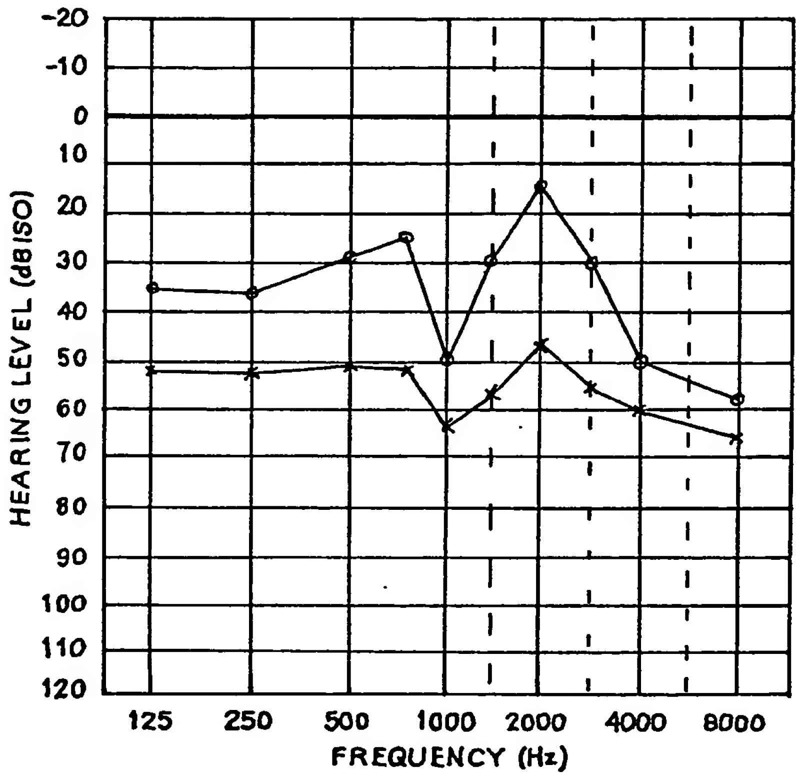
FIG. 8



SPECIFICATION

Compensation of Hearing

This invention relates to a signal processing device for improving a user's hearing ability. In one particular, but not exclusive, application the invention relates to a hearing aid system for improving hearing, e.g. compensating a hearing deficiency.

It is already known to provide, in a hearing aid, a signal processing device comprising a bank of filters each having a different pass band and to scale the amplitude of the output signal from each filter to compensate for a particular individual's hearing deficiency. However, such a known hearing aid does not optimise the use of the filters, especially where the hearing deficiency to be compensated is not uniformly defective over the entire audible spectrum.

According to one aspect of the invention a signal processing device for improving a user's hearing ability comprises signal processing means having an input for processing an electrical signal representative of an auditory signal and an output for supplying a processed electrical signal to transducer means for producing an output auditory signal modified to improve a user's hearing ability, the signal processing means including filter means having a transfer function determined by a plurality of settable parameters and a programmable memory for storing the said parameters of the filter means, and calibration means for determining the said filter parameters to be stored in said programmable memory in dependence on a measured hearing spectrum of a user having the hearing to be improved, the said calibration means optimising the assignation of the available filter parameters to provide an optimised filter transfer function for improvement of the user's hearing.

The invention is primarily intended for use in a hearing aid system for compensating impaired hearing (i.e. a hearing deficiency). In this case a first transducer means is provided for receiving an input auditory signal and supplying an electrical signal to the input of the signal processing means and a second transducer means is provided for connection to the output of the signal processing means. However, the invention is not limited solely to its application in an in-the-ear hearing aid since it also finds application in improving the quality of received sound, e.g. by filtering out unwanted noise in, for example, aircraft communication systems, factory noise situations, the use of stethoscopes and hi-fi systems. Furthermore, for example, the signal processing device may be incorporated in a telephone set for the use of a person having a hearing impairment or deficiency.

According to the invention, the transfer function of the filter means is matched to the hearing spectrum of the user. Conveniently the transfer function is based on the inverse of the hearing spectrum of the user although, to provide the best optimisation of the compensating filter transfer function it will generally be necessary to slightly modify the match with the user's inverse hearing spectrum to compensate for such factors as the variation with frequency of the apparent "equal loudness" for a given volume, the modification required for the very hard of hearing and the inadequacies of transducer means, e.g. in hearing aids.

Preferably the calibration means includes calculating means for assigning the pole and zero locations of the transfer function to minimise the error, preferably the mean squared error, between the inverted hearing spectrum of the user (possibly modified or compensated as mentioned above) and the compensating spectrum of the filter means, in this manner an optimised filter match is obtained since the pole and zero locations can be assigned anywhere in the frequency domain. Thus, for example, with a user having a hearing deficiency only in certain parts of the hearing spectrum, the available poles and zeros can be assigned only to those parts of the hearing spectrum requiring compensation.

A number of techniques are possible for the minimisation of error between the inverted hearing deficiency spectrum (possibly modified or compensated as mentioned above) and the compensating spectrum of the filter means. For Finite Impulse Response filters the principal technique is the Remez Exchange Algorithm and for Infinite Impulse Response filters the principal technique is the minimisation of mean squared error using Fletcher-Powell optimisation.

The programmable memory may be designed so that it is charged only once when the transfer function of the filter means is adapted to the user's hearing deficiency. Preferably, however, the programmable memory is erasable, i.e. an EPROM ("erasable programmable read only memory"). In this latter case the filter means can be periodically "re-programmed", e.g. on the occasion of a further audiometric examination in the case of a hearing aid wearer, to cater for periodic changes in the user's hearing impairment.

The processing means may comprise an analog-to-digital converter for converting the input electrical signal into a digital signal, a digital processing unit incorporating said filter means and a digital-to-analog converter for converting the processed digitial signal into an analog signal. In this case the signal processing in the processing unit is carried out completely digitally. Alternatively, however, the processing means may employ filter means functioning time-discrete and amplitude-analog. Such time-discrete filter means exhibit all the advantages of pure digital filters but no longer require analog-to-digital and digital-to-analog converters. Examples of such time-discrete filters are switched capacitor filters (SCF), bucket brigade filters ("bucket brigade devices"—BBD) and filters with charge-coupled memories ("charge coupled devices"—CCD).

According to a second aspect of the invention a hearing aid system comprises a signal processing

device according to said first aspect of the invention, first transducer means for receiving an input auditory signal and converting it into an electrical signal for supply to said input of the signal processing means, and second transducer means connected to said output of the signal processing means for producing said output auditory signal.

According to a third aspect of the invention a telephone set comprises means for receiving audio signals, electroacoustic transducer means having an input side and an output side for producing output sound signals, a signal processing device and switching means operable to pass said received audio signals to the input side of the electroacoustic transducer means either via the signal processing device so that the transducer means produces compensated output sound signals compensated for a hearing deficiency or so as to by-pass the signal processing means.

Preferably the signal processing device includes programmable signal processing means which can be tailored to compensate a particular user's hearing deficiency. In this case the signal processing means includes a programmable memory which may be charged only once when adapted to the user's hearing deficiency. Preferably, however, the programmable memory is erasable, e.g. an EPROM, to enable the signal processing means to be periodically "re-programmed" to cater for periodic changes in the user's hearing impairment.

Conveniently, the signal processing device comprises signal processing means which are easily removable as a unit from the telephone set. In the preferred case, where the signal processing means is programmable, the programmable unit may be removed from the telephone set and taken to a remote location for adaption to a particular user's hearing impairment. Alternatively, however, there may be a plurality of standard signal processing means each intended for compensating different hearing deficiencies. In this latter case, it is essential for the different signal processing means to be in the form of removable units, the desired unit being chosen to match the particular hearing deficiency to be compensated.

A preferred signal processing device is one according to said first aspect of the invention.

The electroacoustic transducer means may comprise a speaker mounted in a handset (e.g. the normal handset speaker of a telephone set) and/or a further speaker, e.g. a conference speaker. In the case of two speakers being provided, the signal processing means may be switched so as to modify both or only one of the two speakers.

The signal processing means and switching means may be housed together in a handset of the telephone set, together on a base part of the telephone set on which the handset normally rests when not in use, or one in the handset and one in the base part.

The processing means may be designed to improve the hearing of more than one user, e.g. in the case where two or more users with impaired hearing share the same telephone set. In this case, the processing means would conveniently comprise a number of different transfer functions each matched to a different user's hearing impairment. The switching means would then suitably be operable to select the required transfer function for a particular user.

Embodiments of the invention will now be described, by way of example, with reference to the accompanying drawings, in which:—

Figure 1 is a schematic block circuit diagram illustrating the principle of a hearing aid system according to the invention,

Figure 2 is a schematic block diagram of a calibration unit of the hearing aid system shown in Figure 1,

Figures 3 and 4 are schematic block circuit diagrams of two different embodiments of hearing aid for a hearing aid system according to the invention,

Figure 5 is a flow chart illustrating a filter optimisation procedure,

Figure 6 is a graph showing two audiogram plots,

Figure 7 is a graph showing equal loudness curves, and

Figure 8 is a graph showing two further audiogram plots.

Figure 1 shows a block circuit diagram of a hearing aid system comprising a hearing aid 1 and a calibration unit 2 which can be temporarily connected to the hearing aid 1 for the transfer of data thereto. The hearing aid includes a hearing aid microphone 3 as an input transducer, a hearing aid speaker 4 as an output transducer and, connected between the two transducers, a processing unit 5 incorporating a multi-stage filter 6 and a field programmable memory 7.

The filter 6 is designed to be supplied with a plurality of settable or adjustable parameters from the memory 7 which parameters determine the transfer function of the filter. In particular the filter 6 is arranged to have a transfer function which is specifically adapted to compensate for a wearer's particular hearing impairment. Typically the filter 6 is provided with six poles and zeros although this number of poles and zeros is not critical and in more complex filters a greater number, e.g. from 20—30, of poles and zeros can be provided.

The memory 7 for storing parameters of the filter 6 is preferably erasable being in the form of an erasable programmable read only memory (EPROM) or, in particular, an electrically erasable programmable read only memory (EEPROM). With the use of erasable memory, the stored filter parameters can be periodically changed, e.g. on the occasion of an audiometric examination of the hearing aid wearer, so that the transfer function of the filter 6 can be changed to compensate for changes which have occurred in the wearer's hearing impairment.

The parameters to be stored in the memory 7 are computed in the calibration unit 2 shown in schematic block form in Figure 2. The calibration unit 2 includes a signal generator 8 for generating single frequency tones and an amplitude-frequency programming unit 9 connected to the signal generator 8 for enabling the amplitude and frequency of the tone generated by the signal generator 8 to be varied. A headset 10 is connected to the signal generator 8 and a push button 11 is connected to a microprocessor 12, having associated memory and interface components, which is connected to the programming unit 9.

A patient's hearing is tested in the conventional manner used for producing a pure tone (single frequency) threshold audiogram over the audible frequency range, usually in the range 250 Hz to 8000 Hz. In particular the patient's hearing is tested at each of a number of discrete frequencies by gradually increasing the amplitude of the selected tone produced by the signal generator 8 from a low level to a higher level and subsequently gradually reducing the amplitude from a high level to a lower level. The patient is required to actuate the push button 11 when the tone first becomes audible during the period that the amplitude of the tone is gradually increased and again when the tone first becomes inaudible during the period that the amplitude of the tone is gradually decreased. The frequencies at which the patient's hearing is tested are typically at octave intervals although other intervals, e.g. one third octave steps, may be employed. The information relating to the patient's auditory threshold is passed to the microprocessor 10 which averages and stores the hearing threshold characteristic. Based on the measured hearing threshold characteristic, the microprocessor 12 determines the appropriate filter parameters for storage in the memory 7.

The calculation of the filter parameters by the microprocessor 12 is based on the principle of designing a filter 6 having a transfer function which approximates to the inverse hearing spectrum (i.e. the inverse of the measured hearing threshold characteristic) of the patient, although further modifications to prdouce the best optimisation possible of the transfer function may be made by compensating for other factors referred to hereinafter. The microprocessor 12 provides a mathematical operation for assigning the filter parameters, i.e. the pole or zero locations of the filter 6, to minimise the error, preferably the mean squared error, between the inverse hearing spectrum (possibly modified) and the compensating spectrum or transfer function of the filter 6. The use of a mean square error technique for calculating the filter parameters leads to an efficient filter design since the available poles and zeros can be optimally assigned to those parts of the audible frequency range which specifically require hearing aid compensation. Thus, for example, if a patient's hearing is defective only between 2000 Hz and 4000 Hz, all the available poles and zeros can be assigned to that particular frequency range to provide an optimised transfer function for the filter 6.

Once the hearing test using the calibration unit 2 has been completed, the unit 2 is temporarily attached to the hearing aid 1 and the parameters for the optimised filter characteristic are passed via a data link 13 to the memory 7 of the processing unit 5. The calibration unit 2 is disconnected from the "programmed" hearing aid 1 which is then ready to be used by the patient.

A preferred design of hearing aid is shown in Figure 3 in which the processing unit comprises an analog-to-digital converter 20 connected to a hearing aid microphone 21, a digital processor 22 connected to the output of the analog-to-digital converter 20, a programmable memory 25 connected to the processor 22, a digital-to-analog converter 23 connected to the output of the processor 22 and a programmable gain amplifier 24 having inputs connected to the microphone 21, the digital processor 22 and the digital-to-analog converter and an output connected to a hearing aid speaker 65.

The analog-to-digital converter converts the audio signal from the microphone 21 into a set of digital samples. These digital samples are processed according to filtering algorithms, e.g. distributed arithmetic filtering algorithms, incorporated in the digital processor 22 and the pole-zero filter parameters stored in the programmable memory 25. The output of the digital processor 22 is converted in the digital-to-analog converter 23 to an audio signal and is supplied, via the programmable gain amplifier 24, to the speaker 65, the amplifier providing the appropriate broadband gain over the hearing spectrum.

The "filter" of the digital processor 22 may comprise a finite impulse response (FIR) filter or an infinite impulse response (IFIR) filter. If an FIR filter is employed, the microprocessor of the calibration unit (not shown) for determining the filter parameters may employ the Remez Exchange Algorithm for minimising the error between the inverted hearing deficiency spectrum and the transfer function of the filter. If on the other hand, an IFIR filter is employed, the microprocessor of the calibration unit may employ the Fletcher-Powell optimisation for the minimisation of mean squared error.

An alternative design of hearing aid is shown in Figure 4 in which the processing unit comprises a switched capacitor filter 26, a timing generator 27 connected to the filter 26 and a programmable memory 28. The filter parameters are loaded into the memory 28 from a calibration unit (not shown) and are stored in the memory. These filter parameters define the timing waveforms produced by the timing generator 27 which generator determines the characteristics of the switched capacitor filter 26. A hearing aid microphone 29 and a hearing aid speaker 30 are connected to the filter 26.

Thus a hearing aid system is provided in which a hearing aid is programmed to compensate for a wearer's specific hearing deficiency. The compensating filter parameters stored in the hearing aid are calculated so as to optimise the filter transfer function.

There are many ways in which the optimisation of the filter transfer function can be carried out. One such filter optimisation procedure is shown in flow chart form in Figure 5 and is described below.

A weighted error function $E(e^{j\omega})$ is given by

$$E(e^{j\omega})=\hat{W}(e^{j\omega})[\hat{D}(e^{j\omega})-P(e^{j\omega})],$$

where W is a weighting function, D is the desired frequency response and P is the approximating function. The desired response, which is the inverse of the measured hearing spectrum (possibly slightly modified or compensated), is defined on a dense grid of points and an initial guess is made for r+1 extremal frequencies for which an error function magnitude is forced to have the value δ with alternating signs:

$$\hat{W}(e^{j\omega_k})[\hat{D}(e^{j\omega_k})-P(e^{j\omega_k})]=(-1)^k\delta$$

with K=0, 1...r
The value of δ is calculated from

$$\delta=\frac{a_0\hat{D}(e^{j\omega_0})+a_1\hat{D}(e^{j\omega_1})+\ldots+a_r\hat{D}(e^{j\omega_r})}{a_0/\hat{W}(e^{j\omega_0})-a_1/\hat{W}(e^{j\omega_1})+\ldots+(-1)^r a_r/\hat{W}(e^{j\omega_r})}$$

where

$$a_k=\prod_{\substack{i=0\\ \neq k}}^{r}\frac{1}{x_k-x_i}$$

$x_i=\cos\omega_i$
Interpolation of $P(e^{j\omega})$ on the r points $\omega_0, \omega_1\ldots\omega_{r-1}$ is achieved by

$$C_k=\hat{D}(e^{j\omega_k})-(-1)^k\frac{\delta}{\hat{W}(e^{j\omega_k})}\quad k=0, 1\ldots r-1$$

and

$$P(e^{j})=\frac{\sum_{k=0}^{r-1}\left[\left(\frac{\beta_k}{x-x_k}\right)C_k\right]}{\sum_{k=0}^{r-1}\left(\frac{\beta_k}{x-x_k}\right)}$$

with

$$\beta_k=\prod_{\substack{i=0\\ \neq k}}^{r-1}\frac{1}{x_k-x_i}$$

and

$x=\cos\omega$

The filter coefficients are derived from its impulse response using the Discrete Fourier Transform over 2^M equally spaced frequencies, where $2^M \geq N$, with N equal to the total filter delay.

As previously mentioned, the optimised filter transfer function is preferably based on the measured hearing spectrum, e.g. the pure tone audiogram. Two audiogram plots are shown in Figure 6, plot A showing a typical plot for an ear displaying normal hearing and plot B showing a typical plot for an ear displaying impaired hearing. In plot B there is a threshold shift at 4000 Hz to 60 dB so that hearing is impaired in the vital speech frequency band (500 Hz to 4000 Hz). This results in speech being less intelligible and, in particular, sharp consonants ("t" and "s") lose their acuity. Although a threshold shift of 30 dB is not normally sufficient to prevent normal comprehension, with a threshold shift of 60 dB hearing can be considered virtually ineffective.

Although a transfer function having a positive mirror image of the audiogram plot B in Figure 6 would dramatically improve the ear hearing performance, and is intended to fall within the scope of the invention, such a transfer function can be further optimised to enable restoration of virtually normal hearing by taking account of the following factors:

(1) The apparent "equal loudness" for different frequencies, for a given volume, varies as is apparent from the equal loudness plots shown in Figure 7. Thus it may be necessary to boost or amplify the volume at certain frequencies.

(2) The greater the hearing loss or deficiency, the less the peak to peak variations tend to be —see, for example, the two audiogram plots shown in Figure 8. Thus it may be necessary to provide a different "gain factor" after the dB loss has been compensated for.

(3) The inadequacies of the hearing aid transducers may result in non-linear output and accordingly these may need to be compensated.

The software in the hearing aid can be programmed to provide such a further optimised transfer function.

It will be realised that the transfer function of the filter means may be modified in other respects to filter out certain types of unwanted noise. For example any noise generated by the hearing aid microphone could be filtered out by the filter means. Alternatively, or in addition, near stationary, relatively long duration noise (e.g. machinery or vehicular noise) could be detected and filtered out in the processing means. Other typical applications for the invention, where noise can advantageously be electronically filtered out, include electronic filtering ear "muffs" for machine operators, aircraft communications (e.g. for pilots), stethoscope ear pieces for filtering out unwanted "hiss" and hi-fi noise reduction systems.

The invention has been specifically described in relation to a signal processing device for a hearing aid. However the invention also finds application in other fields, e.g. in relation to a telephone set provided with a signal processing device. In this latter case (not shown), a telephone set, typically comprising a handset and a base, is modified by incorporating a signal processing device and a switch operable to pass an audio signal either directly to the speaker (or speakers) of the telephone set or to the speaker(s) via the signal processing means. The signal processing device is preferably as previously described in relation to the hearing aid. However any signal processing device which improves a user's hearing ability may be employed.

The signal processing device and switch may be housed together in a handset of the telephone set, together on a base part of the telephone set on which the handset normally rests when not in use, or one in the handset and one in the base.

The processing device may be designed to improve the hearing of more than one user, e.g. in the case where two or more users with impaired hearing share the same telephone set. In this case, the processing device would conveniently comprise processing means having a number of different transfer functions each matched to a different user's hearing impairment. The switch would then suitably be operable to select the required transfer function for a particular user.

CLAIMS

1. A signal processing device for improving a user's hearing ability comprising signal processing means having an input for processing an electrical signal representative of an auditory signal and an output for supplying a processed electrical signal to transducer means for producing an output auditory signal modified to improve a user's hearing ability, the signal processing means including filter means having a transfer function determined by a plurality of settable parameters and a programmable memory for storing the said parameters of the filter means, and calibration means for determining the said filter parameters to be stored in said programmable memory in dependence on a measured hearing spectrum of a user having the hearing to be improved, the said calibration means optimising the assignation of the available filter parameters to provide an optimised filter transfer function for improvement of the user's hearing.

2. A signal processing device according to claim 1, in which the transfer function of the filter means is matched to the hearing spectrum of the user.

3. A signal processing device according to claim 1 or 2, in which the transfer function of the filter means is based on the inverse of the hearing spectrum of the user.

4. A signal processing device according to any of the preceding claims, in which the filter means comprise a Finite Impulse Response filter.

5. A signal processing device according to any of claims 1 to 3, in which the filter means comprises an Infinite Impulse Response filter.

6. A signal processing device according to any of the preceding claims, in which the calibration means includes calculating means for assigning the pole and zero locations of the transfer function to minimise the error between the inverted hearing spectrum of the user and the compensating spectrum of the filter means.

7. A signal processing device according to claim 6, in which the pole and zero locations are assigned to minimise the mean squared error between the inverted hearing spectrum of the user and the compensating spectrum of the filter means.

8. A signal processing device according to claim 6 when dependent upon claim 4, in which the calculating means makes use of the Remez Exchange Algorithm to minimise said error.

9. A signal processing device according to claim 5 and claim 6 or 7, in which the calculating means makes use of the Fletcher-Powell optimisation to minimise said error.

10. A signal processing device according to any of the preceding claims, in which the programmable memory is erasable.

11. A signal processing device according to any of claims 1 to 3, in which the processing means comprises an analog-to-digital converter for converting the input electrical signal into a digital signal, a

digital processing unit incorporating said filter means and a digital-to-analog converter for converting the processed digital signal into an analog signal.

12. A signal processing device according to any of claims 1 to 3, in which the filter means functions time-discrete and amplitude-analog.

13. A hearing aid system for improving a user's hearing ability comprising a signal processing device according to any of the previous claims, first transducer means for receiving an input auditory signal and converting it into an electrical signal for supply to said input of the signal processing means, and second transducer means connected to said output of the signal processing means for producing said output auditory signal.

14. A hearing aid system substantially as herein described with reference to, and as illustrated in, Figures 1 and 2, Figure 3 or Figure 4 of the accompanying drawings.

15. A telephone set comprising means for receiving audio signals, electroacoustic transducer means having an input side and an output side for producing output sound signals, a signal processing device and switching means operable to pass said received audio signals to the input side of the electroacoustic transducer means either via the signal processing device so that the transducer means produces compensated output sound signals compensated for a hearing deficiency or so as to by-pass the signal processing means.

16. A telephone set according to claim 15, in which the signal processing device includes programmable signal processing means which can be tailored to compensate a particular user's hearing deficiency.

17. A telephone set according to claim 16, in which the signal processing means includes a programmable memory.

18. A telephone set according to claim 15, in which the signal processing device is in accordance with any of claims 1 to 12.

19. A telephone set according to any of claims 15 to 18, in which the transducer means comprises a speaker mounted in a handset and/or a speaker not mounted in the handset.

20. A telephone set according to any of claims 15 to 19, in which the signal processing means and switching means are both provided in a handset of the telephone set, are both provided on a base part of the telephone set on which the handset normally rests when not in use, or one is provided in the handset and one is provided in the base part.

21. A telephone set according to any of claims 15 to 20, in which the processing means comprises a number of different transfer functions each matched to a different user's hearing impairment.

22. A telephone set according to claim 21, in which the switching means is operable to select the required transfer function for a particular user.

Printed for Her Majesty's Stationery Office by Courier Press, Leamington Spa. 6/1987. Demand No. 8991685.
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